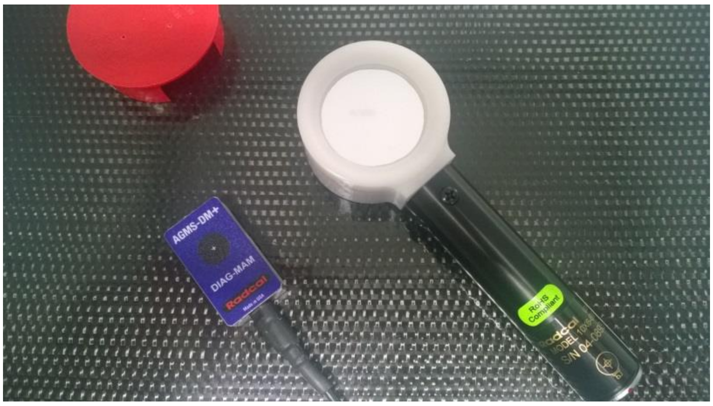
Dosimetry and diagnostic reference level in mammography



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1. Introduction

According to the radiological protection principle of optimization, there is a need to limit the dose from X-ray imaging without impairing the reliability of the diagnosis (ICRP 2007). Radiation doses should be determined at regular basis and compared to diagnostic reference levels. For comparable and reliable results, measurements should be performed using uniform methods and regularly calibrated measuring instruments. Average radiation exposure from mammography should be annually determined in at least one imaging view by calculation or by measurement.

Previously, Entrance Surface Dose (ESD) was generally used to quantify radiation exposure due to mammography. The use of digital equipment in mammography and the larger range of available anode and filter materials have broadened the selection of applied radiation qualities. Radiation exposure in mammography is recommended to be quantified by calculating the Mean Glandular Dose (MGD) (or Average Glandular Dose AGD) (IAEA 2007). MGD estimates radiation exposure in the radiation sensitive glandular tissues and gives a better estimate of risk and detriment than ESD, which only considers the dose on the skin of the patient. The Mean Glandular Dose takes into account breast thickness and glandular consistency as well as parameters of the X-ray spectrum.

2. Diagnostic reference level (DRL)

The International Commission of Radiation Protection (ICRP Publication 103, 2007) states that "one of the principles of optimisation of protection in medical exposures is implemented through the use of diagnostic reference levels (DRLs). The DRL has proven to be an effective tool that aids in optimisation of protection in the medical exposure of patients for diagnostic and interventional procedures". Reference level refers to a predetermined X-ray examination radiation dose level that is not presumed to be exceeded in a procedure performed according to the standards of good practice upon a patient of normal size.

The national DRL values are based on surveys or registries. Values of appropriate DRL quantities from patient examinations are collected from several different health facilities. The ICRP recommends, in general, that the median value of the DRL quantity from each facility is derived from the distribution of the dosimetric data for each type of examination at that facility (ICRP, 2017). Also, the national DRL value is set at the 75th percentile of the distribution among the facilities. For mammography, the DRL quantity can be e.g. air kerma (unit mGy) measured or calculated from patient or phantom data. The most simple approach could be setting DRL values for breasts of 50 ± 5 mm thickness.

Further, DRLs are not aimed for comparing and limiting the exposure in individual cases but merely for comparing the average dose to a group of individuals exposed at conditions of standard good practice. It may be necessary to exceed the reference level in order to gain better image quality. On the other hand, even if the obtained exposure levels are well below the DRL, issues of radiation safety may need optimization for the particular exposure. It is important to ensure that image quality is sufficient for a reliable diagnosis and that at the same time the radiation exposure is not excessive.

In the following, examples are given on how diagnostic reference level for mammography examinations are defined in different European countries:

1) **In Norway**, the diagnostic reference levels for Average Glandular Dose (AGD) in mammography are issued by the National Radiation Protection Agency (NRPA) and defined as follows:

Imaging projection	AGD per exposure (mGy)
CC (cranio caudal)	1,3
MLO (mediolateral oblique)	1,5
Total (per examination, 2 projections)	3,0

The national DRL in Norway has been calculated based on the 75 percentile of the data obtained from local DRLs (Friberg et al. 2015). Further, in the Regulations on Radiation Protection and Use of Radiation it is stated that representative doses must be reviewed regularly. More specifically, it is stated that dose values should be reviewed at least every three years, and more frequently if new equipment, methods or protocols are introduced.

2) **In Finland**, the corresponding values are issued by Radiation and Nuclear Safety Agency and given as Mean Glandular Dose (MGD):

Imaging projection	MGD per exposure (mGy)
CC (cranio caudal)	1,5
MLO or LAT (mediolateral oblique or lateral)	1,5

For all clinics performing mammography, the requirement is that radiation exposure is evaluated at intervals of no more than three years. The individuals for the dose evaluation should be selected so that the compressed breast thickness is 40-60 mm and the average thickness ca. 50 mm. Exposures are determined for the most common examinations with each equipment and at least for one imaging projection for which a reference level has been issued. Values from at least ten patients in a group are either measured or calculated.

3) **In Portugal**, a DRL proposal has been established based on a survey comparing phantom and patient data: DRLs from patient data was 1,5 mGy per projection and for phantom it was 1,2 mGy per projection.

4) **In Switzerland**, there is no DRL as such for mammography exposure. Instead, the maximum average glandular dose levels set for typical breasts simulated with PMMA phantoms are used (Table 1):

Table 1: Dose levels for typical breasts simulated with PMMA (European guidelines, EU 2013)

Thickness of PMMA (mm)	Equivalent breast thickness (mm)	Maximum average glandular dose to equivalent breasts (mGy)	
		Acceptable level	Achievable level
20	21	≤ 1.0	≤ 0.6
30	32	≤ 1.5	≤ 1.0
40	45	≤ 2.0	≤ 1.6
45	53	≤ 2.5	≤ 2.0
50	60	≤ 3.0	≤ 2.4
60	75	≤ 4.5	≤ 3.6
70	90	≤ 6.5	≤ 5.1

3. Determination of mean glandular dose (MGD) for individual patient

3.1. Calculation of MGD

$$\mathbf{MGD = K_i * g * s * c} \qquad [1]$$

- K_i =incident air kerma at patient skin
- g = conversion coefficient for glandular dose (assuming 50% glandularity), accounting for radiation quality (HVL) and breast thickness
- s = coefficient for X-ray spectrum (anode and filter material)
- c = coefficient accounting for glandularity of the breast

In the following, the different factors in the MGD formula will be elucidated

3.2. Incident air kerma (K_i) at patient skin

is determined with air kerma K_a. The distance for the air kerma measurements performed with the ionization chamber (FCD, focus-to-chamber distance) should be proportionated to the correct distance for the breast skin (FSD, focus-to-skin distance) used in the patient exposure settings. The correction is performed with the inverse square law method.

$$\mathbf{K_i = K_a (FCD/FSD)^2} \qquad [2]$$

where

FSD (focus to skin distance)
FCD (focus to chamber distance).

The imaging values are often stored in the image, or, they are visible on the control panel screen after the mammography exposure has been performed. The performance of the equipment measuring the thickness of the breast should be checked and corrected accordingly.

3.3. Air kerma (K_a)

is determined either by

1) measuring it by using ionization chamber with the exposure settings used for patient imaging

$$\mathbf{K_a = \sum(M_1\text{-}M_n)/n * N_{k,Q} * k_{Tp}} \qquad [3]$$

- Measurements (M_1-M_n) with an ionization chamber are performed using the mammography equipment with craniocaudal imaging settings and the compression plate positioned as high as possible in order to avoid scatter radiation. The ionization chamber is placed on top of the breast support plate (directly on top of phantom or breast support plate if the chamber is protected from backscatter) with the reference point 60 mm from the chest wall edge and centered laterally (Figure 1). The temperature- and air coefficient (k_{Tp}) is established and calibration coefficient ($N_{k,Q}$) for the ionization chamber is noted. The k_{Tp} is not required for semiconductors, as their operation is not dependent on pressure or temperature.

$$k_{Tp} = ((273{,}15+T)*101{,}3) / 293{,}15 * p \qquad [4]$$

T= air temperature, p = air pressure (some ionization chambers have inherent correction for temperature and pressure, so it is crucial to be aware of the features of the radiation meter in use).

Figure 1. Measurement of air kerma using ionization chamber on top of phantom (left) and a semiconductor directly on breast support plate (right).

OR,

2) using the equipment's tube output (Y) determined for all applicable radiation qualities, i.e. combinations of anode/filter material and tube voltage range.

Tube output (Y, mGy/mAs) of the X-ray equipment can be calculated using the patient settings in the X-ray equipment: tube voltage and tube loading (Q; mAs) and measured air kerma

$$Y= K_a / Q \qquad [5]$$

- X ray tube output at the reference point is determined for each anode/filter combination in clinical use and for a set of tube voltages and tube loadings which adequately represent the patient exposure parameters used.
- Measurements of tube output for a selected series of voltage settings used in mammography are performed and can be presented graphically (tube output as function of tube voltage for **each distance settings** in the equipment). The curve can be used for interpolation of tube output at non-measured voltages.

3.4. Conversion coefficients

Coefficient g for different HVL and breast thickness
It is essential to know the HVL (half value layer) (mm aluminium, Al) for each radiation quality as this defines the conversion or correction coefficient g. This coefficient is received from appropriate tables where breast and HVL are shown at various thicknesses (Dance et al. 2000; European commission 2013). HVL is the filter thickness that attenuates the air kerma measurement to half of the value obtained without filter. It is determined by first measuring air kerma without filters (air kerma free in air), and then add, on top of the compression plate, an appropriate thickness of Al filters, and air kerma is again measured with the same settings. The thickness of the Al filters is increased until a value has been reached where the air kerma is less than half of the original value measured without filters. The exact value of the half value layer can be interpolated either from an exponential function or by estimating it from a graph where air kerma values are plotted as function of Al thickness.

Coefficient s for anode material and filter material
The coefficient values are given in tables (Dance et al. 2000, Dance et al. 2009, European Commission 2013) for different anode/filter material combinations used in the mammography equipment.

Coefficient c for glandular content
The coefficient for breast glandularity can be obtained from appropriate tables (Dance et al. 2000, European Commission 2013), in which breast and HVL are shown at various thicknesses. Since the glandular content, i.e. the proportion of glandular tissue in the breast, is generally not known for an individual, the c coefficient is given separately for women in age group 50 to 64 and for age group 40 to 49.

In the event that the mammography device is equipped with a display for either ESD or MGD or both, these can be used, provided that the readings are validated regularly.

The dose calculation methods described here are valid for conventional mammography imaging (projection) and can be used for dose calculation of an individual patient. In case of tomosynthesis or 3-D equipment, the required measurements can be performed by halting the equipment in upright position and applying a specific coefficient for different angles (Dance 2011).

4. Determination of MGD using phantom

Using the phantom it is possible to perform

- dose comparison between different mammography equipment,
- quality assurance for AEC performance and image control
- simulation of an average breast in estimation of patient doses and comparison to DRL

The radiation attenuation abilities of a 4.5 cm thick PMMA phantom equals those of an average 5 cm thick breast with equal parts (50%/50%) of adipose and glandular tissues.

It should be noted that phantom based dose estimation does not correspond to an average dose calculated from a group of patients. ***Measurements using phantom is merely an estimate for the dose distribution of patient data*****,** and it cannot be used for comparing functional values in mammography equipment in imaging of different breast types. Patient doses can vary to a great extent, even if the breast thickness would be the same. This is due to differences in breast composition and construction: if the AEC sensor is located near glandular tissue, the dose is bigger than if the sensor would be located near adipose tissue.

In practice, exposure parameters are determined by placing the 45 mm thick PMMA-phantom on the breast support plate. The compression plate is positioned as high as possible. Exposure of the phantom is performed using automatic exposure control (AEC) and the settings are recorded (tube loading (Q(auto), tube voltage, and target/filter combination). It is important that the automatically selected settings are the same as those used in clinical exposures of the standard breast. Determination of air kerma, incident air kerma and the MGD can then be performed using the formulas 3, 2 and 1, respectively. For the phantom approach, the c factor is 1, as the PMMA phantom is considered to represent 50% glandular and 50% adipose tissue.

It should be observed that in the manual settings it is not always possible to use exactly the same tube loading (mAs) settings as those obtained from the automatic exposure using the phantom. In such case, measurements (M) with an ionization chamber are performed with value of the tube loading (Q(man) as close as possible to the one measured with phantom (Q(auto)). Air kerma (K_a(man)) is determined using the above formula ($K_a = (M_1\text{-}M_n/n) * N_{k,Q} * kpT$). Air kerma for the tube loading in the phantom exposure is achieved by

$K_a(auto) = (K_a(man)/Q(man)) * Q(auto)$, K_a(auto)= air kerma for AEC settings in phantom exp.
K_a(man)= air kerma for manual settings
Q(auto)= tube loading for AEC settings
Q(man)= tube loading for manual settings

5. Radiation meters

Determination of air kerma of the mammography equipment should be performed with adequate radiation meters suitable for the energy range of X-ray radiation and intended for measuring mammography devices. Both ionizing chambers and semiconductor meters are used for the purpose. Radiation meters shall be calibrated in accordance to international standards (eg. IEC 61674) and the calibrations shall be traceable to metrological standards. Calibrations are performed regularly, at implementation of a new meter and thereafter at five years intervals at the least. Several radiation qualities are used in the calibration and calibration coefficients are calculated for each quality.

6. Examples of dosimetry calculations

6.1. Measurement of air kerma and tube output

In a mammography equipment with Mo-anode and 25 µm Rh-filter, imaging values 28 kV and 63 mAs are used for measurements with ionization chamber. The values obtained are M_1 = 5,56 mGy, M_2 = 5,59 mGy and M_3 = 5,57 mGy. Air pressure is 99 kPa and air temperature 22 °C. The calibration coefficient for the radiation meter is 1,02.

Air kerma is determined using formula (3): $K_a = \sum(M_1\text{-}M_n)/n * N_{k,Q} * k_{TP}$,

and k_{TP} using formula (4): $k_{TP} = ((273{,}15+T)*101{,}3)) / 293{,}15 * p$
$= ((273{,}15+22)\ °C *101{,}3\ kPa)) / (293{,}15\ °C * 99\ kPa) = 1{,}03$

thus, $K_a = (5{,}56 + 5{,}59 + 5{,}57 / 3)* 1{,}02 * 1{,}03 = 5{,}86$ mGy

Tube output : $Y = K_a / Q = 5{,}86 / 63 = 0{,}093$ mGy/mAs

6.2. Determination of MGD for patient

Mammography is performed using Mo-anode and 25µm Rh-filter, tube voltage 29 kV and tube loading 50 mAs. The age of the patient was 45 years and breast thickness 5,5 cm. The distance between focus and support plate is 63 cm and the distance to ionization chamber is 61 cm. Tube output at 29 kV is 0,132 mGy/mAs (obtained from a tube output curve established for the Mo/Rh combination at different voltage settings, see Figure 2). Determine MGD for the patient.

1. Air kerma at breast surface (**K_i**) is calculated using formulas 2 and 5:

$$\mathbf{K_i = K_a (FCD/FSD)^2 = Y * Q (FCD/FSD)^2 = 0{,}132 * 50 * (61/(63\text{-}5{,}5))^2 = 7{,}43\ mGy}$$

2. Half value layer is determined from an established Mo/Rh curve (not available) at 29 kV: 0,441 mm Al

3. MGD is calculated with formula (1)

$$\mathbf{MGD = K_i * g * s * c} \qquad \mathbf{[1]}$$

Values for g, s, and c are interpolated from the European Guidelines (2013) Tables A5.1, A5.4a, A5.7:
g = 0,208
s= 1,017
c= 1,039

Thus, MGD = 7,43 mGy 0,208 1,017 1,039 = **1,63 mGy**

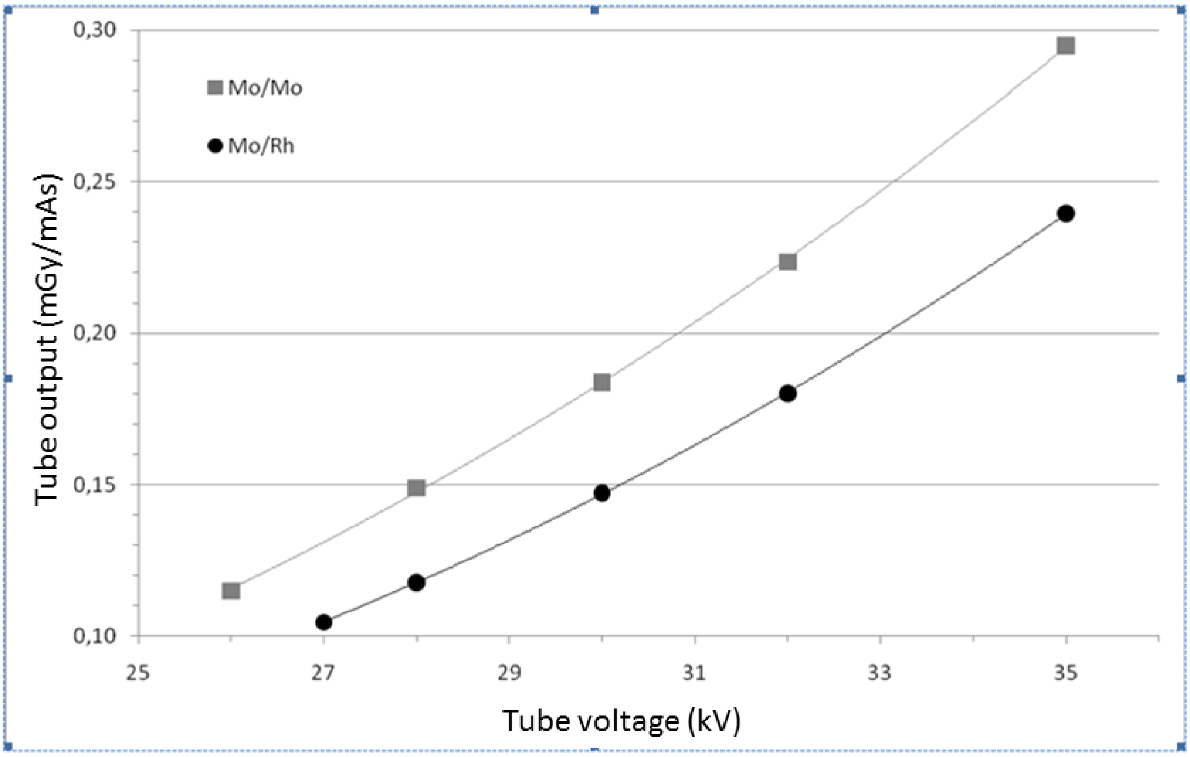

Figure 2. Tube output as function of tube voltage for a mammography equipment with two different filter combinations: 30 µm Mo/Mo and 25 µm Mo/Rh.

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